

# **Muscle Activation Patterns during Linemen Movements in NCAA Football Players**

Undergraduate Honors Thesis

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**Abstract:**

Damage to articular cartilage in the knee can lead to pain and prevent knee functionality. Once damaged, cartilage has limited ability to self-repair, presenting significant danger for progression into severe conditions such as osteoarthritis. Some types of cartilage injuries are known as cartilage defects and are common in athletes, particularly in linemen participating in American football. It is known that excessive forces in the knee can cause cartilage defects; these forces may arise from excessive and repetitive torsional loading of the joint or single external traumas to the knee. However, it is unknown why linemen in particular are so commonly diagnosed with defects. It is possible that activation of certain muscles surrounding the knee could be an important factor in causing high forces in the knee. In this project, NCAA linemen were asked to walk, jog, and perform rush trials. During these movements, 3-D motion data of linemen were collected with the use of a motion capture system and electromyography (EMG) data was collected with surface electrodes. Each movement was broken down into multiple stages and EMG data were then used to calculate a co-contraction index for each stage. Additionally, these motion data were entered into an open-source software, OpenSim, which can be utilized to create subject-specific musculoskeletal simulations of movements. A statistical analysis was run on the results for co-contraction, and it was determined that antagonistic muscles surrounding the knee activate more and for a longer period of time in certain stages of rush and jog as compared to walking. In addition, it was found that, due to model simplifications of the lumbar spine, simulations of motion could not be performed.

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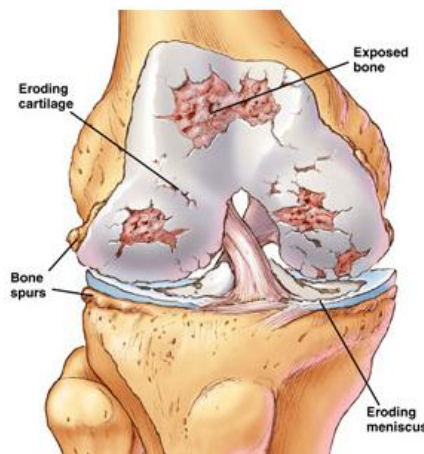
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## Chapter 1: Introduction

Articular cartilage is present in joints throughout the body. In the knee, it acts to support loads applied to the joint and provides a nearly-frictionless surface for joint movement [1].

Articular cartilage consists of an extracellular matrix predominantly made up of water, proteoglycans, and collagens [2]. This matrix consists of only one type of cell, the chondrocyte, which are thinly dispersed throughout the matrix [2]. Extreme loading or twisting of the joint can cause damage to cartilage and may result in death of chondrocytes [2, 3].

If the cartilage is injured, it can dramatically decrease functionality, basic movement, and can change how loads are distributed within the knee [1]. However, blood vessels and nerves are not present in articular cartilage, which can prevent detection of injury and inhibit healing if cartilage is damaged [2]. Certain types of cartilage injuries are referred to as defects, and are often described as “potholes” in cartilage (Figure 1) [4]. Without intervention, cartilage defects can progress into more serious conditions such as osteoarthritis [5].



**Figure 1:** Osteoarthritis of the knee

<http://www.ouh.nhs.uk/hipandknee/information/knee/arthritis.aspx>

Defects in the cartilage of the knee are common, with about one million cases occurring in the United States each year [6]. Athletes are particularly prone to cartilage injury; it has been



suggested that defects occur in 36% of all athletes, a much higher prevalence than in the general population [7]. Joint injury and degeneration are specifically common for athletes competing in sports which subject joints to high impact and torsional loading [3]. American football is considered as one of these high impact sports, and linemen are particularly at risk for developing cartilage defects [3, 8]. In a 15 year study of cartilage injuries in the NFL, about half of the injuries were to linemen [8].

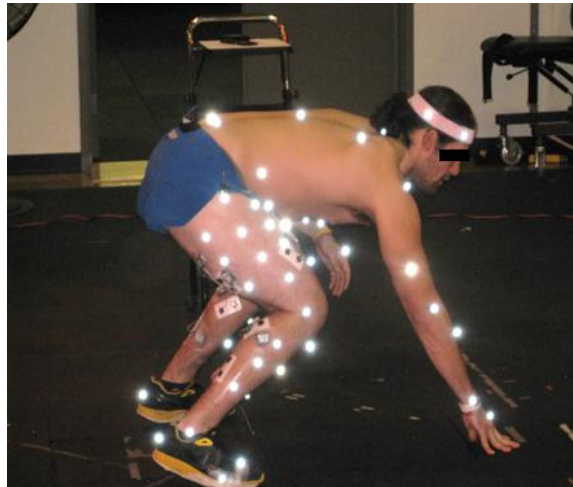
While the exact cause of defects remains unclear, there are certain factors which are known to increase the likelihood of cartilage injury. Articular cartilage may be damaged when large loads are applied to the joint and certain motions can cause higher impacts and greater twisting of the knee joint than others. Movements such as running, jumping, squatting, or cutting will put higher loads on the knee than are experienced while walking [3]. During walking, peak tibiofemoral contact forces are about 2.7-2.8 x BW (body weight) [9, 10]. In comparison, tibiofemoral contact force is about 3.6-4.2 x BW for jogging [11, 12] and 4.3 x BW for squatting [13]. Studies have shown that contact stresses between 4 and 9 MPa are commonly experienced in healthy articular cartilage during daily activities and do not cause observable cartilage injury [3]. However, single impact articular cartilage contact stresses greater than 25 MPa result in significant chondrocyte death, and therefore substantial cartilage damage [14, 15]. An example of this type of a single impact may be a direct hit to the joint with a football helmet [3].

In addition to single impact loads, excessive and repetitive loading may also injure cartilage; studies show that there is a strong association between excessive, repetitive mechanical stresses and progression of cartilage degeneration [3, 16, 17]. One study showed that repetitive

loading of cartilage results in fatigue failure, which shows an accumulation of chondrocyte death with repetitive contact stresses as low as 6 MPa [18].

Initial injury to the cartilage in the knee may leave the joint susceptible to additional damages. For example, the location and manner in which forces are distributed in the joint can be changed due to injury or avoidance of pain [19]; this may overload certain areas of the cartilage that are not accustomed to experiencing high loading, resulting in a higher rate of defect progression [19]. Moreover, once a defect occurs, the “pothole” can act as a stress concentration producing enhanced stress around the defect, causing progression by damage to nearby healthy tissue [20].

Squats, vertical jumps, jogging, and coming out of a three-point stance (Figure 2) into forward, backward, and angled-back movements are all common for linemen.



**Figure 2: The three point stance**

The muscles in the leg are responsible for a person being able to perform these movements, but the forces from these muscles also contribute to the loads seen at the cartilage in the knee. Forces from the muscles surrounding the knee can either prevent cartilage damage or accelerate it. In some cases, muscles contract to absorb much of the energy which is applied to the knee

from high impacts or twisting motions [3]. In this way, the muscles help protect and stabilize articular surfaces in the knee [3, 21]. In other cases, sets of muscles on both sides of the knee joint may contract simultaneously, or co-contract, in order to gain greater stability or “stiffen” the knee. While some level of co-contraction of the hamstrings and quadriceps is obligatory to stabilize the knee in healthy individuals, persons with quadriceps weakness oftentimes experience a prolonged co-contraction of these muscles which increases the contact stress in the articular cartilage of the knee [22, 23]. In this case, co-activation can reduce the ability for the knee to absorb shock effectively [22].

Due to the increased prevalence of cartilage injury in linemen, it is possible that patterns of muscle activation during lineman-specific movements are generating large forces in the knee that may be partially responsible for the high rate of cartilage defects. The purpose of this project was to determine which lower extremity muscles were being used during different movements which linemen commonly perform in order to study the relationship between muscle function and loads at the knee in this at risk population.

## **1.1 Focus of Thesis**

The goal of this research was to study how football linemen use their muscles during different movements in order to determine possible reasons for why linemen are at an elevated risk for knee cartilage injuries. The movements which were studied were gait, jogging, and rush drills. One of the emphases of this research was to analyze the aforementioned movements in order to characterize a co-contraction index of muscles surrounding the knee. Additionally, simulations of these movements were attempted in order to determine the forces that the muscles exert at the knee and patterns of muscle activations. Together, the co-contraction index and

patterns of muscle activations can help characterize possible causes of cartilage knee injury during different movements which are common to linemen.

## **1.2 Significance of Research**

It has been well documented that football linemen experience a high risk for cartilage injury [3, 8]. The muscles surrounding the knee play a key role in retaining stability, aiding mobility, and absorbing forces applied to the joint during different movements [3, 21, 24, 25]. Muscle contraction, specifically muscle co-contraction, increases forces at the knee joint and can lead to cartilage damage [22, 23]. If the patterns of muscle activation and the forces that the muscles exert at the knee during common linemen movements can be characterized, it may be possible to determine a possible reasoning for why linemen are so susceptible to knee cartilage injury.

The simulations which were done in this project were, to the best of our knowledge, the first attempts of simulations of the biomechanics of football linemen's movements. Additionally, this project is an extension of another study done in Ohio State's Neuromuscular Biomechanics Lab which looked at the kinematics and kinetics of linemen's movements. Due to the progressive nature of cartilage defects, it is important to detect, prevent, and manage these injuries early in an athlete's career. By studying the muscles surrounding the leg and the kinematics involved during linemen-specific movements, it may provide needed insight as to possible causes for the high incidence of cartilage injury among football linemen.

## **1.3 Overview of Thesis**

This thesis consists of 6 chapters. Chapter 2 describes how motion data and EMG data were collected and analyzed as well as how co-contraction was calculated. Chapter 3 gives results and statistical analyses of the co-contraction indices for rush drills, as well as walking and

jogging. Chapter 4 discusses the use of Opensim to perform simulations of movement as well as the possible reasons for the failure of these simulations. Chapter 5 gives preliminary results, investigates the failure of simulations with a simple hand calculation, and helps to prove possible reasons for why the simulation failed. The final chapter, chapter 6, provides discussions, conclusions, and future applications of the results found in this thesis.

## **Chapter 2: Methods**

### **2.1 Introduction**

This project was an addition to an ongoing, Institutional Review Board approved study sponsored by the NFL Charities which recruited NCAA linemen before and after a football season in order to study defect progression in the articular cartilage of their knees. High resolution magnetic resonance images (MRI) of both of each subject's knees were taken before and after a season of play in order to determine presence of cartilage defects. However, the linemen in this study had no known history of injury or surgery to either of their knees before participation in this study. Pre-season data was previously collected by other members of the lab, primarily Becky Lathrop, and work for this thesis involved collecting and processing post-season data. Motion capture data from before and after the season was then processed in order to examine muscle forces and patterns of muscle activations at the knee during different movements.

### **2.2 Data Collection and Motion Capture**

Reflective markers were placed on the subject's skin according to the Point Cluster Technique (PCT) [26] (Figure 3). Knee motion was captured at 300 Hz using ten Vicon MX-F40 cameras to track relative movement of reflective markers. During all movements, EMG data (TeleMyo DTS System, Noraxon Scottsdale, AZ) were collected at 600 Hz or 1500 Hz (due to experimental error, (Table 1). Electromyography (EMG) electrodes were placed on both the left and the right legs over the following muscles: rectus femoris, vastus lateralis, vastus medialis, biceps femoris, semitendinosus, lateral and medial gastrocnemius, and soleus (Figure 3).

**Table 1:** EMG collection rate for each subject

Subject	EMG Collection Rate
01_F02	600 Hz
02_F02	600 Hz
06_F02	1500 Hz
15_B01	1500 Hz
17_F02	1500 Hz
23_F02	1500 Hz
24_F02	1500 Hz



**Figure 3:** The Point Cluster Technique (PCT) with reflective markers. The white rectangular markers on the upper leg show where EMG electrodes are placed on subjects in order to study their lower extremity muscle activations [from Rebecca Lathrop, Neuromuscular Biomechanics Lab].

## 2.3 Testing Procedures

The subjects were asked to execute a maximum voluntary isometric contraction for each of the aforementioned muscles (Table 2). The athletes then performed a number of non-linemen-specific movements including walking, jogging, and vertical jumping, as well as linemen specific movements including pass, pull, and rush blocks. Six force plates in the floor recorded the reaction forces from the ground during each movement. Meanwhile, these motions were captured using ten Vicon cameras which track 3-D motion of the reflective markers. In addition to dynamic trials, static trials and hip range of motion trials were also recorded.

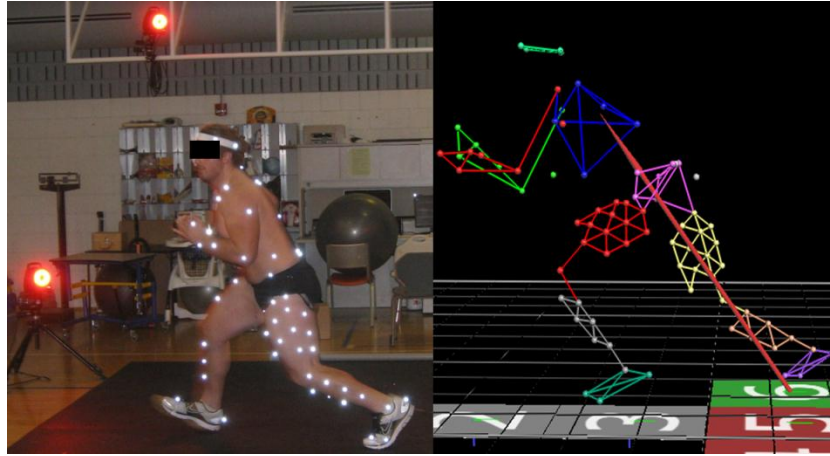
**Table 2: Movements to achieve MVIC values for different muscles**

<b>MVIC Trials</b>	
<b>Muscle</b>	<b>How MVIC was achieved</b>
<i>Rectus Femoris</i>	Subject sits in chair. Chair strap around thigh. Subject attempts to pull knee up.
<i>Vasti</i> (lateral and medial vastus)	Subject sits in chair facing the wall. Strap from wall to ankle. Subject kicks out.
<i>Hamstrings</i> (biceps femoris, semitendinosus)	Subject sits in chair facing the wall. Strap from wall to ankle. Subject bends knee.
<i>Plantarflexors</i> (lateral and medial gastroc, soleus)	Subject sits on floor with back against wall. Strap from wall to foot. Subject points toe.
Note: For some subjects, no straps were available, so subject's motions were resisted by hand.	

## **2.4 Data Analyses**

The motion data that was collected with the use of Vicon infrared cameras and force plates was then implemented into a program known as Vicon Nexus (Figure 4). With custom MATLAB scripts, hip joint centers were determined from the hip range of motion trials using the star-arc approach [27]. Additionally, the hip range of motion and static trials were used to determine anatomical coordinate frames for the femur and tibia. Custom MATLAB scripts were then used to calculate knee flexion and extension angles between anatomical reference frames.





**Figure 4:** Vicon cameras track relative movement of markers (left) and bring it into Vicon Nexus (right) as 3-D motion (the ground reaction force is the red arrow coming from force plate 6 on the ground)

## 2.5 Calculation of the Co-Contraction Index (CI)

For this thesis, it was determined that only three of the captured movements would be analyzed and compared. Specifically, these movements included walking, jogging, and rush drills. Walking and jogging were chosen because these are movements which any football player would perform, thus, giving a basis for comparison to movements only linemen perform. Linemen rush drills were chosen because this movement is considered to be a characteristic movement which linemen perform often and which is similar to other linemen drills such as pull and pass blocks.

Raw EMG data from these movements were high-pass filtered with a fourth-order, phase-corrected Butterworth filter with a cutoff frequency of 10 Hz. A linear envelope was created by full-wave rectification of the EMG data and smoothing with an eighth-order, phase-corrected low-pass Butterworth filter with a 25 Hz cutoff frequency. This smoothed data was then normalized for each muscle to the maximum activation found in either the MVIC or dynamic trials (gait, jog, and rush). Co-contraction of antagonistic muscles surrounding the knee was

characterized by the co-contraction index (CI) as described by Schmitt and Rudolph [23, 28].

The following equation was used to find the CI [23, 28]:

**Equation 1**

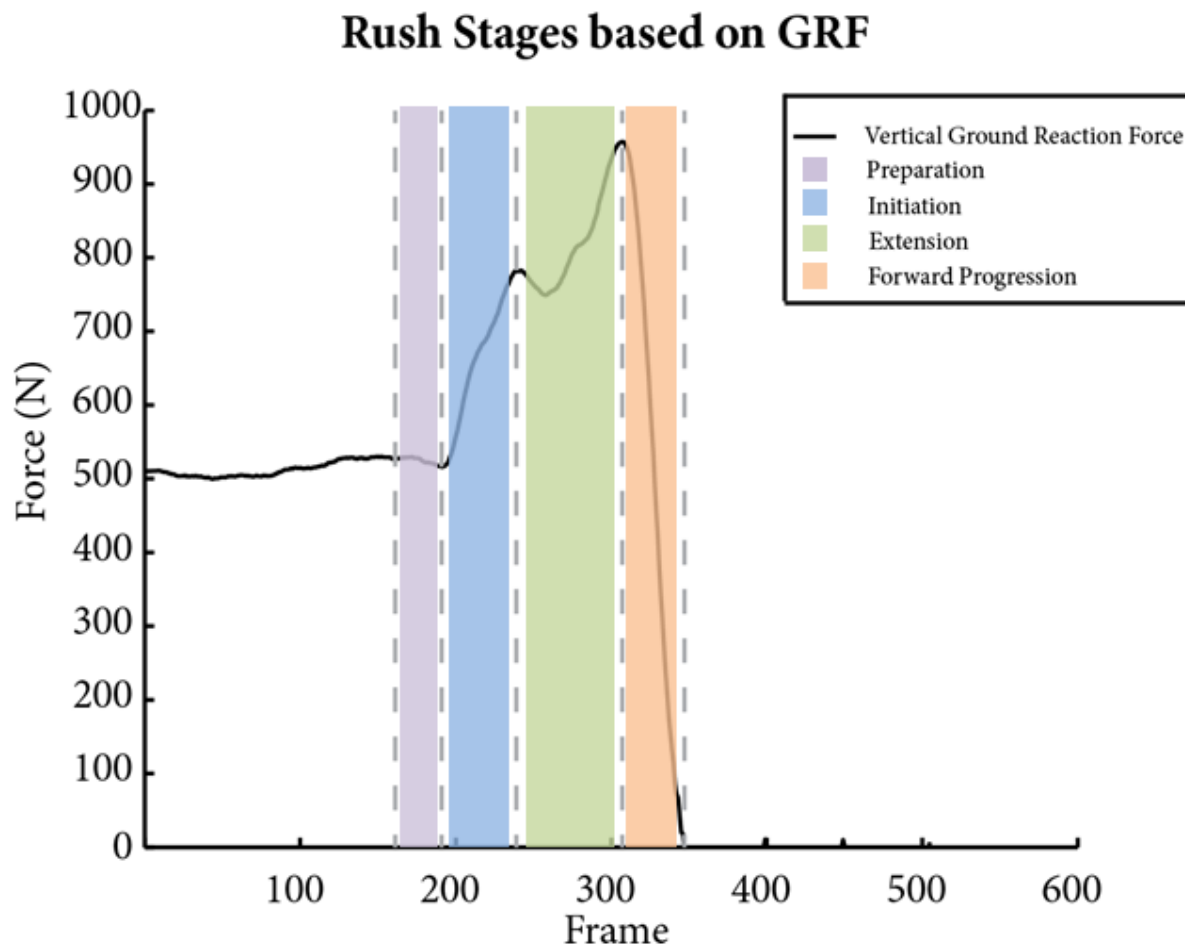
$$Co - contraction\ Index = \frac{\left[ \sum_{i=1}^{51} \frac{lowerEMGi}{higherEMGi} (lowerEMGi + higherEMGi) \right]}{51}$$

From Equation 1, *lowerEMGi* and *higherEMGi* signify which muscle has a lower or higher (respectively) normalized EMG value on a point-by-point basis. The only difference between Equation 1 and the equation that Schmitt used is that these data could only be broken into 51 data points rather than 100 due to the slower EMG collection rate in this study. From Equation 1, a large CI is indicative of a high level of activation of the two specified antagonistic muscles over a large amount of time [28]. In order to utilize Equation 1, each activity was broken up into sub-intervals. However, EMG data for each movement were time normalized so that timing of events within each interval coincided across different subjects.

The stance phase of the gait and jog trials were identified using force plate data. Within the stance phase, gait and jog trials were broken into three intervals: preparation, weight acceptance, and mid-stance. As described by Schmitt, the preparation stage was defined from 100 ms prior to initial contact until initial contact, weight acceptance was defined from initial contact to peak knee flexion angle (during early stance), and midstance was defined from peak knee flexion angle to peak knee flexion angle (during early stance) [23]. As described in section 2.3, the peak knee flexion and extension angles were found using inverse dynamics and inverse kinematics calculations in Vicon Nexus.

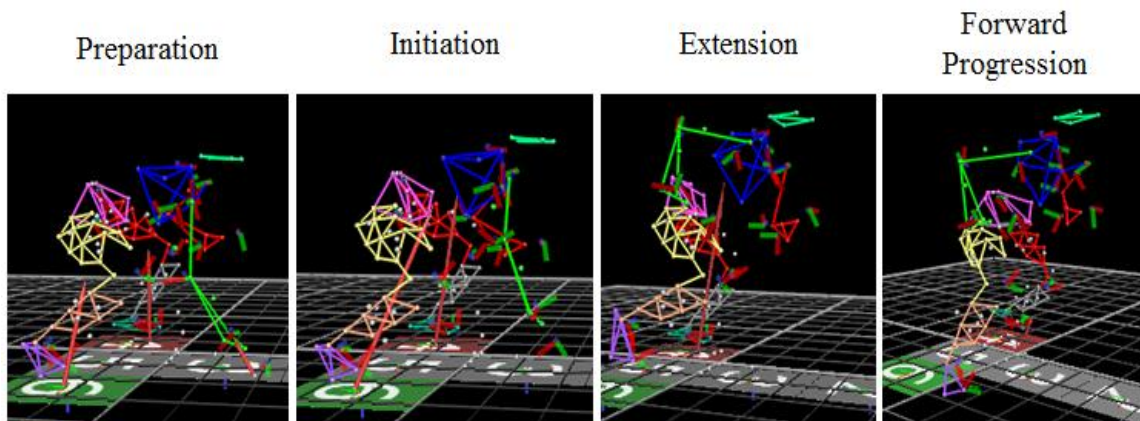
The rush drills are unique in that they did not have previously defined intervals from the literature. Instead, the rush drills were broken up into four intervals which the authors defined: preparation, initiation, extension, and forward progression (Figure 5). For the rush trials, these

stages were chosen based off of the characteristic pattern (for all subjects) of the vertical ground reaction force (GRF) curve (Figure 5) for the leg which remained on the force plates the longest. The leg of the foot which remained on the force plates the longest will be referred to in this thesis as the push-off leg. During rush drills, it is common for linemen to remain in the three point stance (two feet on the ground, crouched forward with one hand on the ground) without moving for a certain amount of time (until the ball is hiked during a game), and then begin to move forward to rush the quarterback. The authors feel that the four stages defined here have effectively captured these movement patterns for linemen rush drills.



**Figure 5:** Characteristic vertical ground reaction force data for the push-off leg of a rush trial. This shows how stages were defined for rush.

For rush drills, preparation was defined as 100 ms prior to movement; movement on the ground reaction force curve appeared as a local minimum in force. Initiation began at initial movement and went until the first local maxima in vertical ground reaction force data (Figure 5). In the initiation stage, the lineman's hand comes off of the ground (Figure 6). Extension begins at the first local maxima in the vertical ground reaction force curve (Figure 5) and continues until the second local maxima in the vertical ground reaction force data (Figure 5). During the extension stage, one of the lineman's feet comes off of the ground (the foot of the non-push-off leg), and the knee of the push-off leg begins to extend (Figure 6). Forward progression begins at the second local maxima in force data (Figure 5) and continues until the push-off foot leaves the force plates. During forward progression, linemen begin to move in the forward direction, away from their initial position and the push-off foot leaves the ground at the end of this stage (Figure 6).



**Figure 6:** The stages of rush as seen in Vicon Nexus

Once all of the activities were broken into their respective stages, co-contraction was calculated between different pairs of antagonistic muscles for each activity. Once the push-off leg was determined for each subject during the rush trial, this leg was determined as the leg of

interest for the gait and jog trials as well, and co-contraction was only calculated for this leg (Table 3). The authors were interested specifically in co-contraction between antagonistic muscles which were on the same side of the leg (both medial or both lateral). For this reason, co-contraction was calculated between the following muscle pairs: vastus lateralis and biceps femoris, vastus medialis and semitendinosus, vastus lateralis and lateral gastrocnemius, and vastus medialis and medial gastrocnemius.

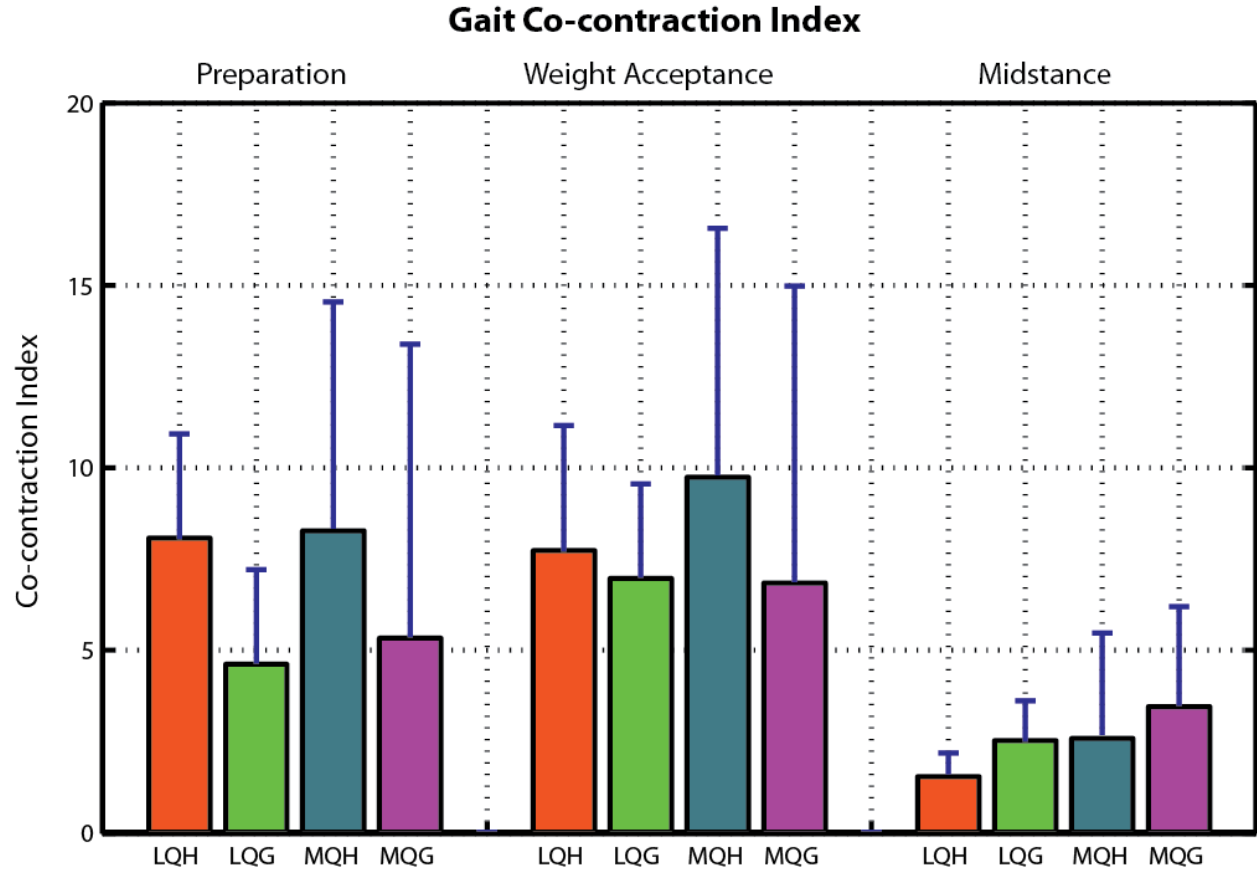
**Table 3:** Push-off leg for each subject

<b>Subject</b>	<b>Push-off leg</b>
01_F02	Left
02_F02	Left
06_F02	Left
15_B01	Right
17_F02	Left
23_F02	Left
24_F02	Right

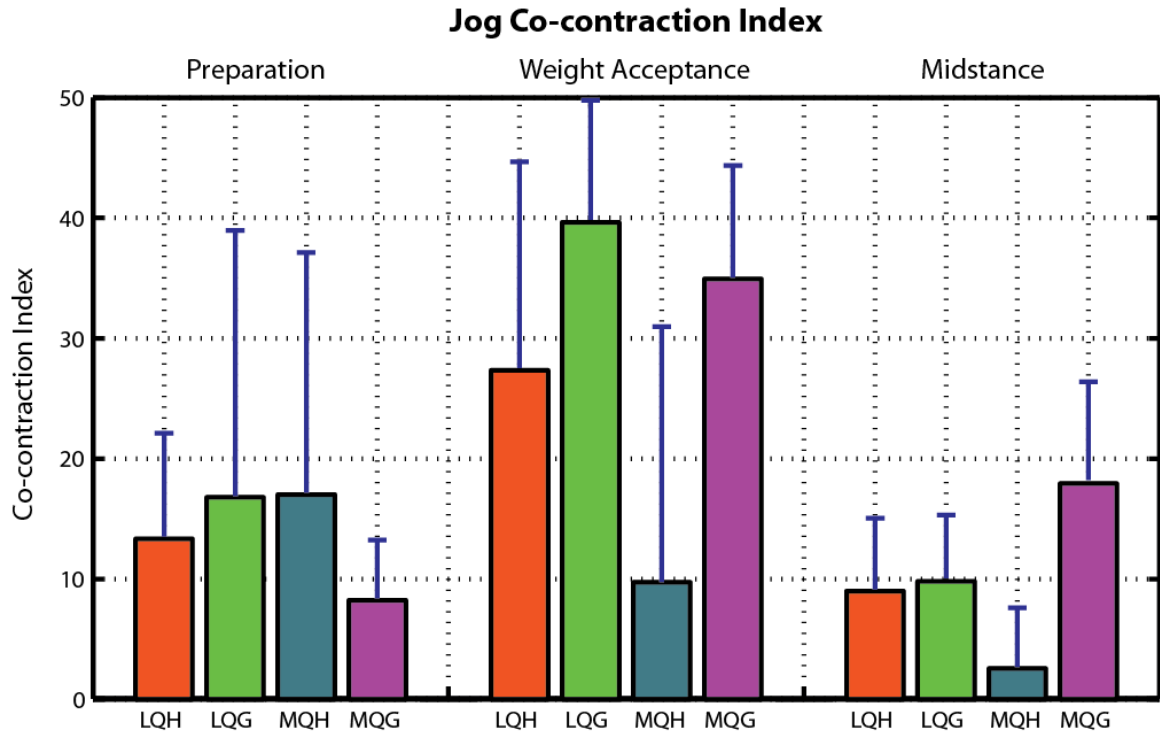
## **Chapter 3: Co-Contraction Results**

### **3.1 Co-contraction during different movements**

The co-contraction index (Equation 1) was calculated for seven subjects for the preparation, weight acceptance, and midstance phases of gait and jog and for the preparation, initiation, extension, and forward progression stages of rush. An average and standard deviation were taken across all seven subjects and all stages of each movement. From the results of the co-contraction index, which can be seen in Figure 7, Figure 8, and Figure 9, it was determined that there was a large range of standard deviations across subjects (ranging from a minimum standard deviation of 0.6513 for the vastus lateralis-biceps femoris during the weight acceptance phase of gait to a maximum standard deviation of 30.5028 for the vastus lateralis-gastrocnemius during the extension phase of rush). In order to compare co-contraction between different muscles, stages, and activities, a statistical analysis was needed.

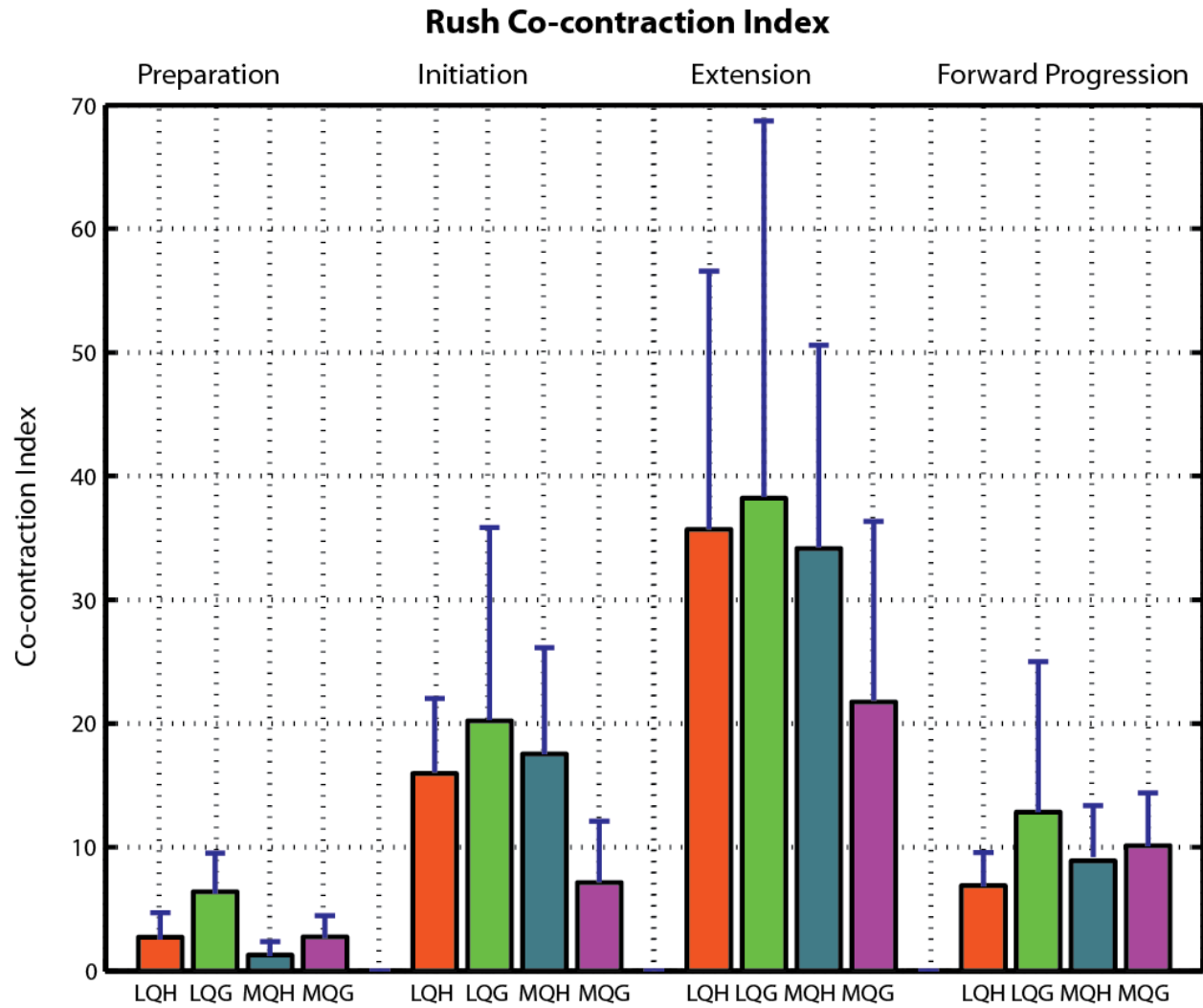


**Figure 7:** Co-contraction during the preparation, weight acceptance, and midstance stages of gait for the push-off leg used in gait. Each co-contraction value represents an average over seven subjects. LQH represents co-contraction between the vastus lateralis and biceps femoris. LQG represents co-contraction between the vastus lateralis and lateral gastrocnemius. MQH represents co-contraction between the vastus medialis and semitendinosus. MQG represents co-contraction between the vastus medialis and medial gastrocnemius. Error bars represent one standard deviation above and below the mean value.



**Figure 8:** Co-contraction during the preparation, weight acceptance, and midstance stages of jog for the push-off leg used in gait. Each co-contraction value represents an average over seven subjects. LQH represents co-contraction between the vastus lateralis and biceps femoris. LQG represents co-contraction between the vastus lateralis and lateral gastrocnemius. MQH represents co-contraction between the vastus medialis and semitendinosus. MQG represents co-contraction between the vastus medialis and medial gastrocnemius. Error bars represent one standard deviation above and below the mean value.





**Figure 9:** Co-contraction during the preparation, initiation, extension, and forward progression stages of rush for the push-off leg used in gait. Each co-contraction value represents an average over seven subjects. LQH represents co-contraction between the vastus lateralis and biceps femoris. LQG represents co-contraction between the vastus lateralis and lateral gastrocnemius. MQH represents co-contraction between the vastus medialis and semitendinosus. MQG represents co-contraction between the vastus medialis and medial gastrocnemius. Error bars represent one standard deviation above and below the mean value.

### 3.2 Repeated measures ANOVA

A statistical analysis of the co-contraction results was performed using Minitab 16 Statistical Software (Minitab Inc., State College, PA). First, a two-way repeated measures ANOVA with Tukey pairwise-comparison was done on only the rush data in order to determine relative trends between stage (preparation, initiation, extension, forward progression), co-contraction muscle-pair (LQH, LQG, MQH, MQG), and the co-contraction value. From this

analysis, it was determined that stage, regardless of which pairs of muscles were co-contracting, had a significant effect on co-contraction value ( $p < 0.05$ ). From the Tukey pairwise comparison, it was established that the value of co-contraction during the extension stage was significantly different, and larger, than in the preparation stage. It was also found that the initiation and progression stages have similar values of co-contraction, as do the forward progression and preparation stages. The specific group of antagonistic muscle pairs used to calculate co-contraction was not found to be significant.

Next, a one-way repeated measures ANOVA and Tukey pairwise comparison were done individually for each stage of rush in order to determine if the muscle pairs compared for co-contraction were significant within a single stage. Significance ( $p = 0.001$ ) was only found within the preparation stage, where LQG co-contraction was found to be significantly different, and larger, than all other types of co-contraction.

To determine the effects of activity and antagonistic muscle pair and their interaction on the co-contraction index, we used a two-way repeated measures ANOVA with post-hoc Tukey pairwise comparison. The activity data were simplified to include only the phases of gait, jog, and rush which generally displayed the largest mean values of co-contraction across the seven subjects. Therefore, this comparison included only the extension phase of rush, the weight acceptance phase of jog, and the weight acceptance phase of gait (Figure 7, Figure 8, Figure 9). Activity significantly influences the magnitude of co-contraction ( $p < 0.001$ ), however, the antagonistic muscle pair does not ( $p = 0.439$ ). The interaction between activity and antagonistic muscle pair was not significant ( $p = 0.439$ ). Comparison between activities showed that the magnitude of co-contraction in the weight acceptance phase of gait was significantly less than for

the weight acceptance phase of jog or the extension phase of rush (both  $p < 0.001$ ). However, jog and rush activities were not significantly different ( $p = 0.926$ ).

## **Chapter 4: Creating Simulations of Movement**

### **4.1 Why use simulations?**

The co-contraction index, patterns of muscle activations, and joint reaction force can all help characterize possible causes of cartilage knee injury during different movements. Studies have shown that when the muscles co-contract, the contact stress at the cartilage increases [22, 23]. From EMG data which was collected during different movements, we can see which muscles are activated and how much they are being activated throughout that movement. From calculating co-contraction index for gait, jog, and rush drills, we can compare the degree to which antagonistic muscles are activating simultaneously. By comparing co-contraction with the same scale across different activities, it is possible to see if one type movement causes the muscles to co-contract more than another. Now that we know which muscles surrounding the knee are co-contracting and how much they are co-contracting, simulations of these movements can be done to estimate how this is affecting the reaction force at the joint. In vivo measurements of joint reaction force would be a very invasive procedure; instead, by using the ground reaction force and kinematic data that have already been collected in Vicon, we can create simulations of movement which allow us to estimate joint reaction forces. These simulations can be done with the results of the static optimization routine in an open-source software known as OpenSim.

### **4.2 Creating a Subject Specific Model**

With the use of OpenSim, subject-specific, dynamic simulations of the athletes' movements can be performed [29]. OpenSim may be utilized to create a subject-specific musculoskeletal simulation which was scaled based on relative distances between pairs of reflective markers taken from the motion capture system as well as the mass of the subject. The OpenSim inverse-kinematics tool can be used to match the model in a pose that best fits the

motion capture data at each instance in time. The inverse-dynamics tool can then be used to calculate joint moments from movement and external reaction force data. The static optimization routine in OpenSim can also be used to estimate what forces the muscles must be exerting in order to perform each given movement and to determine patterns of muscle activation [29]. With static optimization information, we can calculate the reaction force at the knee joint.

#### **4.3 Simulations with the Delp 2392 Torso and Lower Extremity Model**

In order to perform a “proof of concept”, preliminary simulations were run with the use of a torso and lower extremity model developed by Scott Delp [29]. The model consisted of 23 degrees of freedom and 92 muscle-tendon actuators [29] (Figure 10). The isometric strength of the muscles in the model are based off of the strength of a young, adult male and the joint between the pelvis and torso is modeled as a ball-and-socket joint [29].



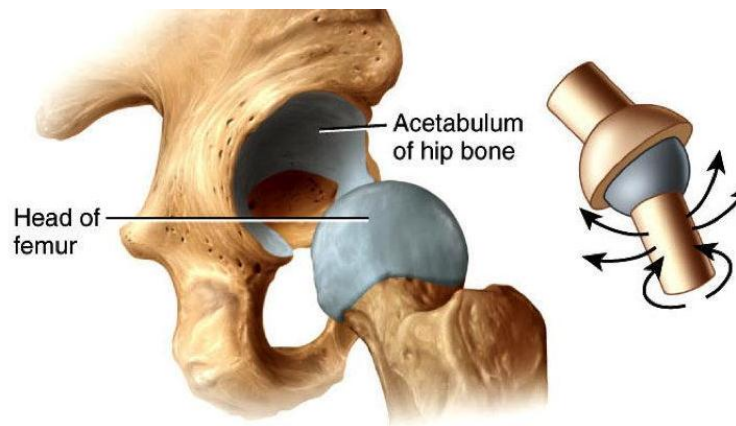
**Figure 10:** Delp’s 92 muscle model of the lower extremity [29]. The red bands signify representations of muscles in OpenSim.

Preliminary simulations were run for the right leg of a single subject during gait, a linemen rush drill, and a vertical jump. Inverse kinematics and inverse dynamics were run for

these movements of interest. However, static optimization could only be run successfully for the walking trial. Since static optimization could not be run for rush or vertical jump trials, joint reaction forces could also not be determined.

#### 4.4 Explanation of Simulation Failure

Simulations for the linemen rush drill and vertical jump trials were not able to be run through static optimization with the Delp 2392 model. The authors hypothesized that there were several possible reasons for why the linemen vertical jump and rush trials did not work with this model, but chose to focus specifically on the rush trial. Part of the issue may be due to the fact that the Delp 2392 model simplified the joint between the torso and pelvis as a ball and socket joint; this basically modeled the torso as a rigid body connected to the lower extremity by a single joint with three degrees of freedom (Figure 11).

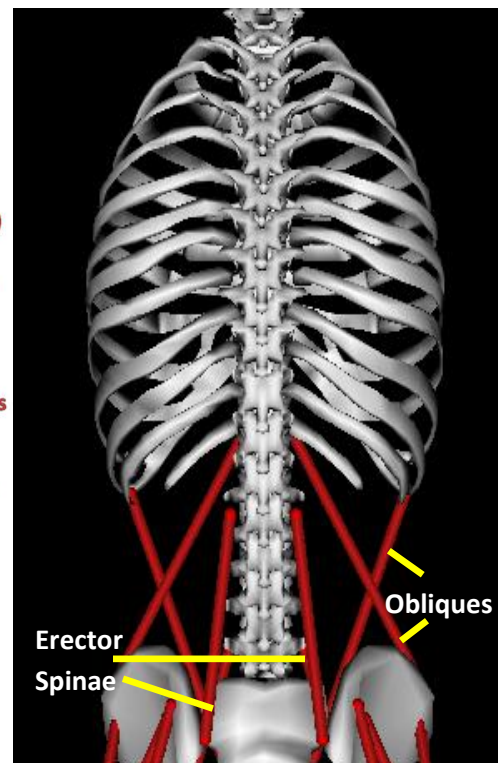
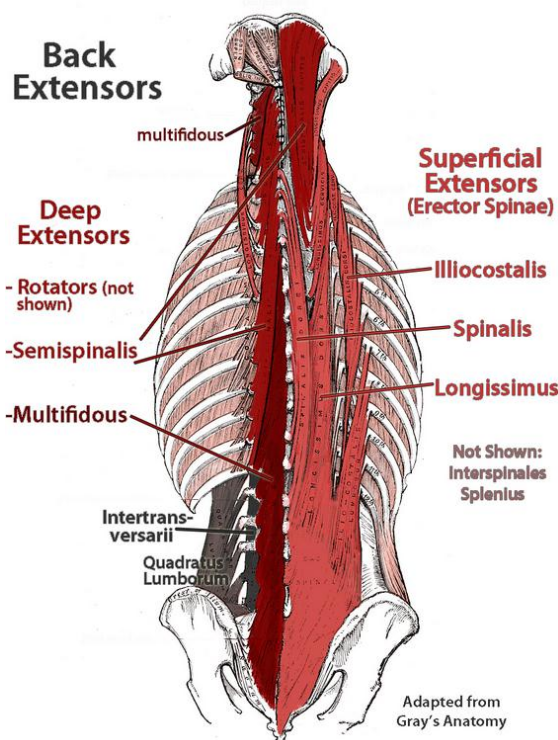


**Figure 11:** Modeling the hip as a ball and socket joint with three degrees of freedom

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In general, during rushing movements, the subject came out of a three-point stance into an upright, running position (as described in Chapter 2). Coming out of a crouched position requires significant lumbar movement, where the recruitment of muscles in the back is vital. With the torso modeled as a rigid body, the lumbar vertebrae are rigid and therefore cannot bend

with respect to each other. Additionally, the lower back muscles were simplified as only the left and right erector spinae and the abdomen muscles were simplified to the interior and exterior obliques (Figure 12). Lumbar movement does not require significant force from the abdomen muscles, so it was assumed that these muscles did not significantly contribute to lumbar motion. This simplification therefore makes it so that significant lumbar torque that is generated during a rush drill is modeled with only two muscles (left and right erector spinae). The authors therefore theorized that static optimization may have failed for rush drills because large back torques would result in extremely large muscle forces being needed for the left and right erector spinae during this movement.



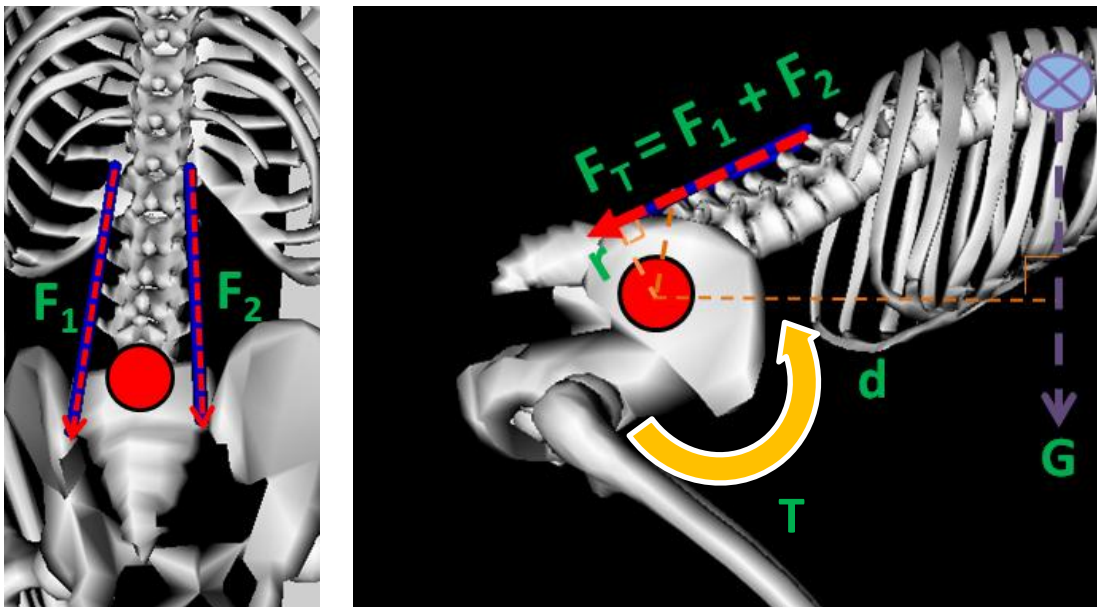
(A)

(B)

**Figure 12:** (A) The actual muscular make-up of the lower back includes many different muscles which can support lumbar movement. (B) The lumbar spine as is simplified in the Delp 2392 model to only include two muscle approximations: the left and right erector spinae, and two abdomen muscles: the interior and exterior obliques.

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To test this theory in OpenSim, seven subjects were scaled, kinematic data was derived using the inverse kinematics routine, and moments were calculated using the inverse dynamics tool [29]. In particular, the lumbar extension moment found from inverse dynamics was important to analyze due to the large torque that is generated from coming out of the three point stance. As a simplification, it was assumed that the right and left erector spinae were the only significant muscles supporting lumbar movement (the forces from the obliques were ignored). With this assumption, a complicated static optimization problem can be simplified so that it is possible to solve for the forces that these muscle must be exerting for the given torque (Figure 13).





(A)

(B)

**Figure 13:** Deriving the internal lumbar extension moment. (A) Top view of the lower back and pelvis. (B) Side view of the pelvis and lower back.

To physically understand the internal lumbar extension moment, it has been simplified to the static case in Figure 13. In Figure 13 A, the blue lines are representations of the left and right erector spinae muscles in OpenSim. The red circle represents the center of rotation for the left and right erector spinae muscles. The dashed red arrows represent the forces which these muscles exert while contracting. In Figure 13 B, the center of mass of the subject's torso segment and the gravity force are in purple and act at a distance  $d$  from the center of rotation. The contractile forces from the left and right erector spinae muscles have been resolved into a single force,  $F_T$  which is a perpendicular distance  $r$  from the center of rotation. These forces and perpendicular distances are what cause the torque,  $T$  about the center of rotation. In this case,  $T$  represents the internal lumbar extension moment, which is equal to the torque caused by gravity. From Figure 13 it should also be noted that because the center of mass is located at a large distance from the center of rotation, the torque due to gravity will tend to be large. This may be another reason why the model fails.

The equation for torque in this example is:

## Equation 2

$$T = (F_T * r) = (G * d)$$

$$T = \text{external flexion moment } (N \cdot m) = \text{internal extension moment } (N \cdot m)$$

$$F_T = \text{muscle force } (N)$$

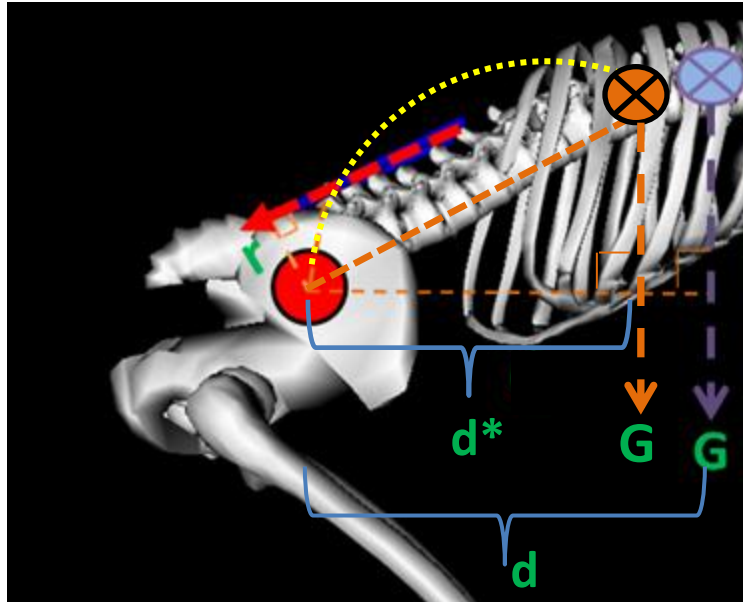
$$r = \text{muscle moment arm } (m)$$

$$G = \text{gravity force } (N) = 9.81 \left( \frac{m}{s^2} \right) * \text{mass of torso segment } (kg)$$

$$d = \text{gravity moment arm (m)}$$

The above parameters were found in OpenSim using results from inverse kinematics and inverse dynamics for each subject. Inverse dynamics results were used to find the maximum lumbar extension moment in Newton · meters and at what time in the trial this occurred at. Then the lumbar extension angle (degrees) was determined from inverse kinematics, at the time in the trial when maximum lumbar extension moment occurred. At this lumbar extension angle, the moment arm (meters) could be determined. With the moment arm and torque known, the forces of the lower back muscles  $F_T$ , can then be solved.

It is important to note that the Delp 2392 model has a rigid back, and therefore relative movement of the vertebrae is not allowed. In reality, when a lineman is crouched in a three point stance, their spine is slightly curved, not rigidly straight. If the back were more curved, the value of  $d$  from Equation 2 would be smaller than if the back were rigid (Figure 14). If  $d$  decreases, the value for muscle force,  $F_T$  will then also decrease, and the internal lumbar extension moment would be smaller. Therefore, it is possible that Delp's model could be giving an artificially large value for the moment arm for the gravity torque, which is then causing an artificially large value for muscle force because of the fact that the vertebrae of the lower back cannot move independently of one another.

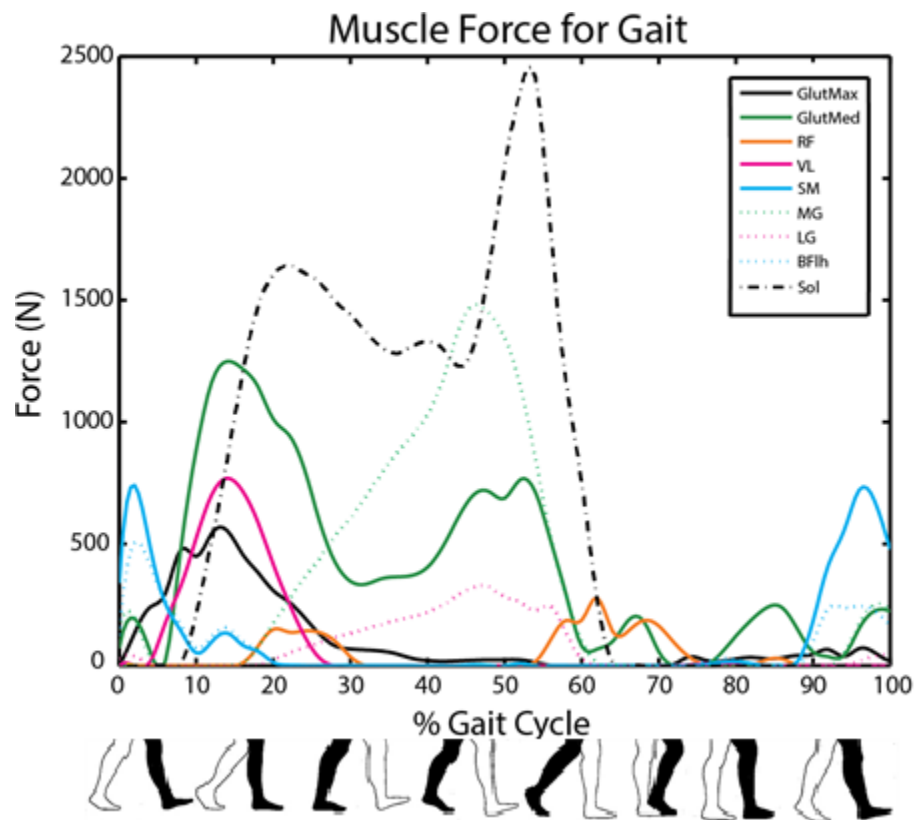


**Figure 14:** The effect of a non-rigid back on gravity moment arm. When the lower back is no longer rigid (yellow dashed line), the value for the moment arm of gravity torque will decrease (  $d^* < d$  ).

## Chapter 5: Simulation Results

### 5.1 Preliminary results

Preliminary simulations were run for the right leg of a single subject during gait, a linemen rush drill, and a vertical jump. The results for muscle forces during a walking trial can be seen in Figure 11. From these data, the largest muscle forces during walking were in the calf muscles (soleus, medial and lateral gastrocnemius) and the glutes (gluteus medius). The results from the muscle forces for a single walking trial were reasonable and similar to those found in the literature [30, 31]



**Figure 15:** Dominant lower extremity muscle forces during a full gait cycle from the right leg (leg pictured in black) of one representative subject

Although preliminary simulations were run for walking trials, they could not be run for vertical jump or rush drills, as the simulations failed. Specifically, the simulations failed for static optimization; this simulation failure for rush drills in particular was further investigated.

## 5.2 Investigating simulation failure

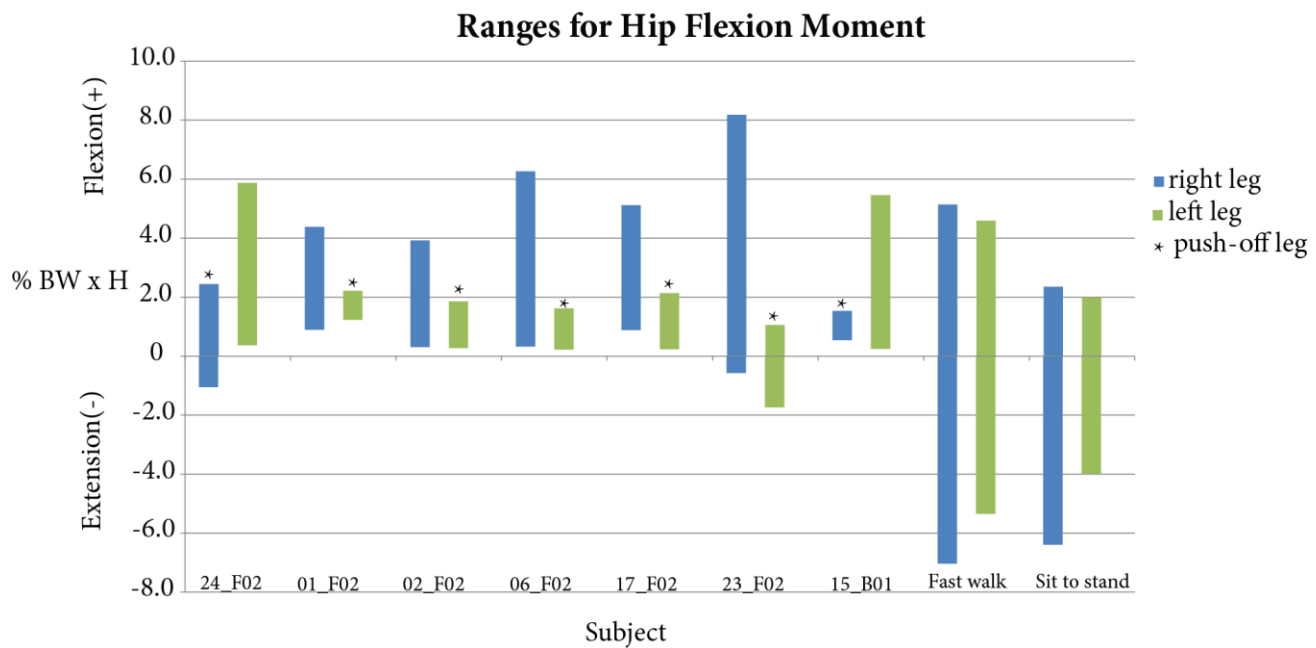
Normally, the static optimization routine in Opensim will calculate individual muscle forces required to perform a specific movement, however, in the case of linemen rush trials, static optimization failed for all seven subjects in this study. In the case where static optimization failed, a simple hand calculation (Equation 2) was done to solve for the forces the lower back muscles exerted in order to perform a rush trial. The co-contraction index calculation uses Equation 1, as described in Chapter 2 of this thesis. From this method, the following results were found for lower back muscle force during rush drills:

**Table 4: Calculation of required muscle force for the given lumbar extension moment**

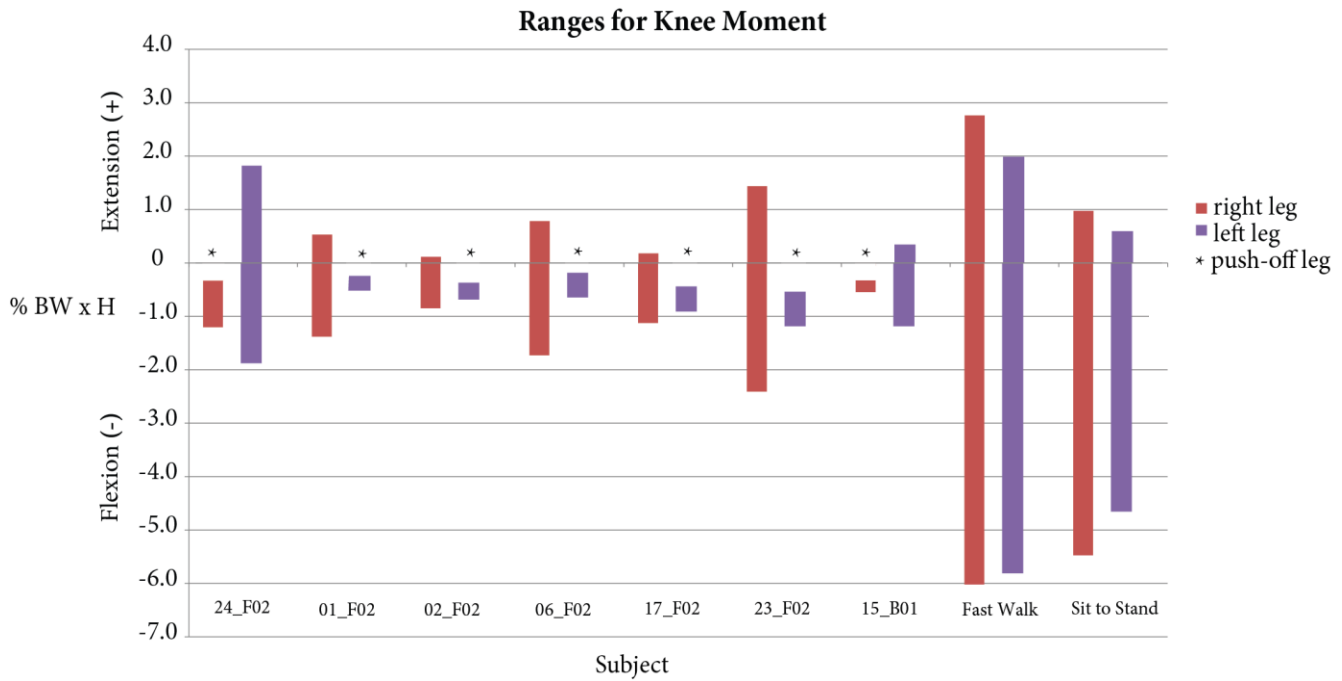
Subject Name	Maximum lumbar extension moment (N·m)	Lumbar extension (angle in degrees)	Moment arm (m)	Combined erector spinae force (N)	Force for each muscle (N)	Amount exceeds maximum isometric force (N)
15_B01	396.411	-69.498	0.002	198,205.500	99,102.750	96,602.750
01_F02	297.640	-34.646	0.026	11,447.692	5,723.846	3,223.846
02_F02	406.354	-73.678	0.001	406,354.000	203,177.000	200,677.000
06_F02	297.269	-69.548	0.002	148,634.500	74,317.250	71,817.250
17_F02	388.499	-75.304	0.002	194,249.500	97,124.750	94,624.750
23_F02	351.690	-40.402	0.024	14,653.750	7,326.875	4,826.875
24_F02	293.117	-40.829	0.019	15,427.211	7,713.605	5,213.605

As can be seen in Table 4, the forces required by the left and right erector spinae muscles were extremely large in order to allow for the given lumbar extension moment. In OpenSim, the maximum isometric force for the erector spinae muscles is set to 2500 N. The forces required for this movement vastly surpass the about of force these muscles can physically produce.

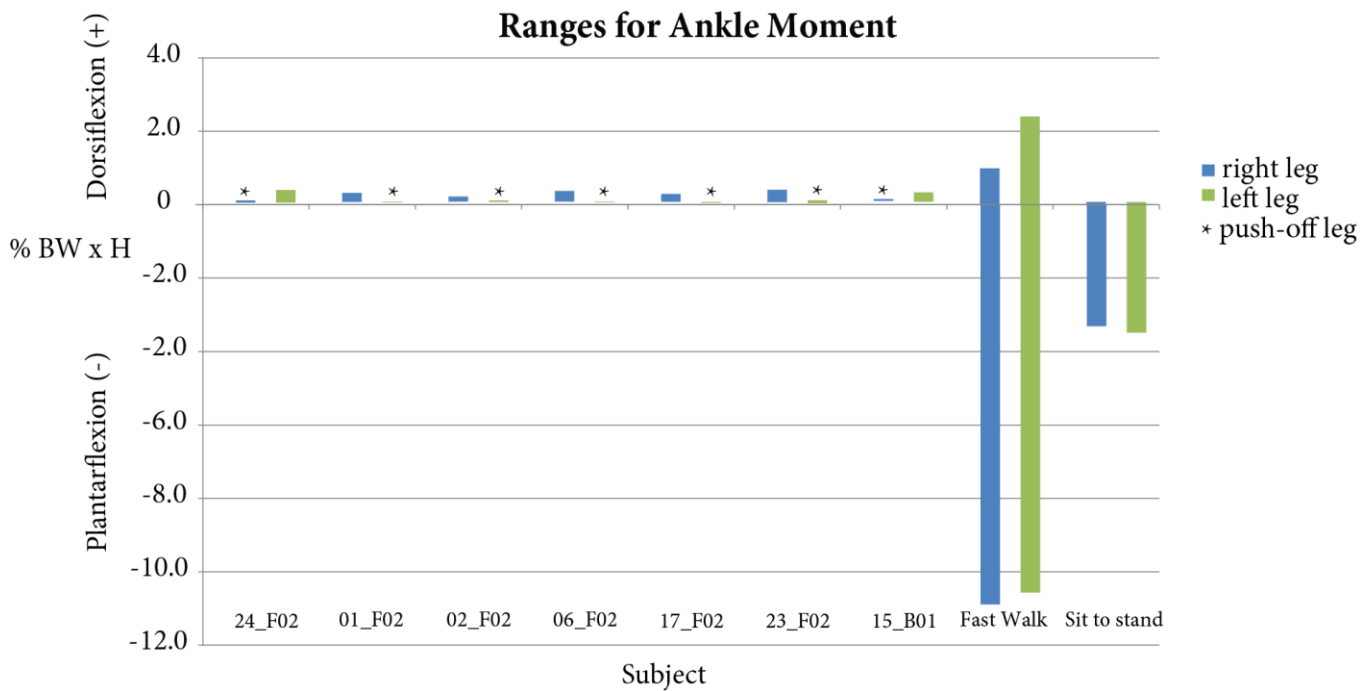
In order to determine if the back muscles were the only reason for the simulation failure, moments about the hip, knee, and ankle during rush drills were compared to sit to stand and fast walk trials (unpublished data from Elena Caruthers, NMBL).



**Figure 16:** Ranges for hip flexion moment from the beginning of the initiation stage to the end of the extension stages of rush.



**Figure 17:** Ranges for knee moment from the beginning of the initiation stage to the end of the extension stages of rush.



**Figure 18:** Ranges for ankle moment from the beginning of the initiation stage to the end of the extension stages of rush.

From the data in Figure 16, it was determined that the ranges for hip flexion moment during rush drills were larger than the ranges for hip flexion during a fast walk and going from a seated position to a standing position (sit to stand).

The knee flexion angle range during rush drills was either extremely close to, or within the range of the knee flexion angle for fast walking and sit to stand movements (Figure 17). Additionally, the knee angle moment for the push-off leg during rush trials was much less than for the other leg which has time to straighten out.

The range of ankle angle moments during rush drills was very small compared to the ranges for a fast walk and sit to stand trial (Figure 18).



## **Chapter 6: Discussion and Conclusions**

### **6.1 Discussion**

It is known that football linemen are particularly prone to cartilage defects, however it is currently unclear which factors are increasing this population's risk for injury [8]. Excessive loads are known to increase the likelihood of injury by increasing contact stresses at the cartilage of the knee [3]. Contact stresses are known to be higher during running, jumping, and squatting than during walking [9-13]. Additionally, when sets of antagonistic muscles co-contract over prolonged periods, contact stresses in the articular cartilage of the knee are increased [22, 23]. Therefore, it is possible that if football linemen are excessively co-contracting the muscles surrounding their knees during specific movements, this may be a factor that elevates this population's risk for experiencing cartilage defects.

From the results of this research, it was determined that linemen are co-contracting more during certain stages within jog and rush trials than in gait trials. Therefore, it is possible that performing these movements may be causing large contact forces in the articular cartilage of the knee. However, because the current musculoskeletal model (Delp 2392) that was used in OpenSim [29] could not be run through static optimization, it was not possible to calculate joint reaction forces in OpenSim. In order to be better-able to characterize factors that are increasing linemen's risk for cartilage defects, issues with the model used in this thesis for simulations were addressed.

From the results of calculating individual muscle force during rush trials for the erector spinae muscles, it was found that the required muscle force greatly exceeded the maximum isometric force that these muscles can produce. Therefore, according to the lumbar extension

moments found using the inverse dynamics routine, this movement is physically impossible for the muscles to support, and it makes sense that the model failed in static optimization.

It is possible that there were issues with other muscles other than erector spinae which may have resulted in the failure of static optimization. In OpenSim, static optimization uses output kinematics and joint moments in order to solve for individual muscles forces; therefore, to check if the model is failing elsewhere, the moments at the hip, knee, and ankle were also investigated.

As previously mentioned, the ranges for hip knee flexion moment during rush drills were larger than the ranges for hip flexion during a fast walk or going from a seated position to standing. A potential reason for this could be that, physically the hip is much more flexed when coming out of a three point stance than while walking. Also, in the sit to stand trial, the hip is not flexed as much as in a rush drill. From this comparison for hip flexion moment in different movements, the authors believe that these moments from OpenSim for rush drills are within reasonable ranges.

Additionally, it was found that the knee moment for rush trials was near to the range of knee moments for fast walking and sit to stand movements. However, typically the knee is more bent during sit to stand than it is during rush, which may be a possible reason for why the knee moment during sit to stand was slightly larger than during rush trials. It was also found that the range for the ankle moment during rush drills was much smaller than for a fast walk or sit to stand. During gait, from heel strike to toe-off there is significant dorsiflexion and planar flexion of the ankle. In a sit to stand trial the ankle angle is very dependent on how bent the knee is when

the feet are planted. During rush drills, the ankle angle did not change much, so the fact that the ankle moment did not change significantly was not surprising.

While moments for the hip, knee, and ankle are believed to be within reasonable ranges, the calculated lumbar flexion moment was larger than what we believed to be physically possible. It was therefore determined that the Delp 2392 torso and lower extremity model [29] was insufficient for the linemen rushing motions and that another model which more accurately represents the muscles in the lower back and allows for relative movement between vertebrae in the back is needed. Due to the fact that static optimization could not be run, it is not possible to use OpenSim to solve for muscle forces or joint contact forces. With only the co-contraction being characterized, it is difficult to infer if the joint contact force in the knee is actually higher and increasing the risk of cartilage defects during rush trials versus jogging or walking.

## **6.2 Contributions**

It is known that football linemen experience a high risk for cartilage injury [3, 8], but it is not currently known what specifically places this population at a higher risk for injury [1, 2]. Muscle contraction, specifically muscle co-contraction, increases forces at the knee joint and can lead to cartilage damage [8, 9]. By studying patterns of muscle activation involved during linemen's movements, EMG data were used to calculate co-contraction indices during gait, jog, and rush trials. It was determined that linemen co-contract more during certain stages of rush and jog trials than during gait. This research also originally aimed to calculate joint reaction forces with the use of OpenSim; however, instead, specific issues with the model that was used were characterized. It was determined that in order to simulate linemen rush movements, a model which can better represent the lumbar spine is needed. With the co-contraction

characterized during different movements, it shows that rush and jog movements may elevate risk for cartilage injury in this population.

### **6.3 Future Work**

From this thesis, it was determined that linemen rush drills cannot be simulated with Delp's 2392 musculoskeletal model [29] in OpenSim. From these analyses, the authors believe that a new model must be developed which has a better estimation of the lumbar spine. A Ph.D. student in the NMBL lab at Ohio State, Elena Caruthers, has begun developing a new, more detailed model. Rather than using the Delp 2392 model for the lower extremity, Caruthers plans to integrate complex models for the lower extremity, lumbar spine, arms, and neck from various models which already exist. Specifically, a lumbar spine model which allows for relative movement between vertebrae is preferable, as this will give a better estimation of movements which have significant lumbar motion.

Once this model is developed, it is the hope that it may be used for movements that have significant lumbar motion; with this new model it may be possible to simulate linemen's movements. With the use of simulations, individual muscle forces can be calculated and patterns of muscle activations can be studied. Other movements which are specific to linemen, including pull blocks and pass blocks may then also be simulated. With the new model, it may be possible to calculate joint contact forces for many different linemen movements. If joint contact force and co-contraction can be characterized during different linemen movements, this may help point to significant factors that are increasing linemen's risk for cartilage injury.

### **6.4 Summary**

The co-contraction indices for gait, jog, and linemen rush movements were characterized. It was determined that the co-contraction for muscles surrounding the knee is not significantly

different for jogging than it was for rush trials. Nevertheless, linemen did co-contract more during certain stages of jogging and rush trials than they did during gait trials. It was also determined that, in a rush trial, linemen tend to have similar co-contraction during the initiation and forward progression stages. However, co-contraction was found to be significantly different during the extension and preparation stages. Additionally, it was determined that a musculoskeletal model with a better representation of the lumbar spine must be used to calculate individual muscle forces and joint reaction forces in movements with significant lumbar motion. This research serves as an important first step in determining which factors are causing football linemen to be more susceptible to cartilage injury.

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